

**DESIGN AND CONSTRUCTION OF A COMPACT MULTI-CHAMBER TISSUE
EQUIVALENT PROPORTIONAL COUNTER**

A Thesis

by

TEMEKA TAPLIN

Submitted to the Office of Graduate Studies of
Texas A&M University
in partial fulfillment of the requirements for the degree of

MASTER OF SCIENCE

December 2005

Major Subject: Health Physics

**DESIGN AND CONSTRUCTION OF A COMPACT MULTI-CHAMBER TISSUE
EQUIVALENT PROPORTIONAL COUNTER**

A Thesis

by

TEMEKA TAPLIN

Submitted to the Office of Graduate Studies of
Texas A&M University
in partial fulfillment of the requirements for the degree of

MASTER OF SCIENCE

Approved by:

Chair of Committee,
Committee Members,

Head of Department,

Leslie Braby
Ian Hamilton
James White
William Burchill

December 2005

Major Subject: Health Physics

ABSTRACT

Design and Construction of a Compact Multi-Chamber Tissue Equivalent Proportional Counter. (December 2005)

Temeka Taplin, B.S., Southern University

Chair of Advisory Committee: Dr. Leslie Braby

This project was designed to determine the feasibility of constructing a multi-chamber proportional counter. A multi-chamber detector is designed to increase the total surface area which will increase the number of radiation interactions that occur per unit dose. Surface area can be changed without changing the detector volume by subdividing the active volume into several smaller volumes that can then be used as mini detectors whose data can be summed and used to determine the absorbed dose. This will allow the total surface area to remain the same as that of the more common 12.5 cm (5 in.) spherical detector and a decreased total volume resulting in a more compact detector design. However, subdividing those volumes causes problems with electric field fringing at the ends of the mini detectors. In order to correct this, guard ring and field tube designs which operate at a lower voltage than the detector cathode were tested. Results from this study showed that the field tube design provided the best overall resolution but it only outperformed the other designs by a maximum of 5%. However the field tube design doubles the length of the detector which would result in a larger overall detector package. The performance of the single and double ring configurations was suitable for

radiation monitoring applications. These findings show that it is feasible to use an array of subdivided detector volumes instead of a spherical detector.

ACKNOWLEDGEMENTS

I would like to thank my committee chair, Dr. Braby, for his guidance throughout the course of my research. I would also like to thank the members of my committee, Dr. White and Dr. Hamilton, for their assistance throughout my research. Thanks to Joe Snook, Tony Restivo and everyone at the machine shop who helped me with this project. I would also like to extend my gratitude to the faculty, staff and many friends that have assisted me during my stay at Texas A&M University. Lastly, I want to thank my family for believing in me and giving me their constant support.

TABLE OF CONTENTS

	Page
ABSTRACT	iii
ACKNOWLEDGEMENTS	v
TABLE OF CONTENTS	vi
LIST OF FIGURES	vii
LIST OF TABLES	viii
 CHAPTER	
I INTRODUCTION.....	1
II BACKGROUND.....	2
Radiation Protection	2
Lineal Energy Transfer (LET).....	3
Microdosimetry	5
Radiation Detection	7
Detectors.....	9
Multi-Chamber Proportional Counters.....	12
III METHODS AND MATERIALS	16
IV RESULTS AND DISCUSSION	25
V CONCLUSIONS	27
REFERENCES	31
APPENDIX	33
VITA	39

LIST OF FIGURES

FIGURE		Page
1	A Representation of High and Low LET Particle Tracks	4
2	Experimental Multi-Chamber Proportional Counter.....	14
3	Test Detector Cavity and Surfaces	17
4	Design of Guard Rings	18
5	Assembly View of Detector Cavity	21
6	Detector Cavity and Attachments	22
7	Design of Voltage Divider Circuit	23
8	Exploded View of Proposed Multi-Chamber Tissue Equivalent Proportional Counter	28
9	Top View of Tissue Equivalent Plate.....	29

LIST OF TABLES

TABLE		Page
1	LET Assumptions.....	6
2	Radii and Voltage Ratios.....	24
3	Resolutions	25

CHAPTER I

INTRODUCTION

The purpose of this project is to design and build a prototype detector that will serve as the basis for a system that will have the counting efficiency of a 12.5 cm spherical tissue equivalent proportional counter but will be much more compact. In order to achieve this, a multi-chamber detector is proposed. Design options utilizing guard rings or field tubes to correct distortion of the electric field that would result in poor pulse height resolution were tested. This prototype detector allows the determination of which of three field correcting techniques is best for this design. The results will provide the basis for the design of a multidimensional array of detectors. The objective of this project is to produce a detector with good counting statistics that will be much smaller than the conventional handheld 12.5 cm (5 in.) spherical detectors that are currently sold for monitoring low dose-rate mixed field exposures.

CHAPTER II

BACKGROUND

Radiation Protection

People are exposed to radiation constantly throughout their daily activities by either radiation in the atmosphere, in our homes, or from the earth itself. Much of this radiation is the result of elements decaying to achieve nuclear stability. In order to gain this stability the nucleus emits particles that have an energy distribution that is characteristic of a nuclear transition. The emitted particles (alpha and beta) are then able to transfer their energy to matter which can cause several different effects depending on the properties of the interacting material. Unstable elements can also release excess “nuclear” energy without an associated charged particle; this is known as gamma radiation. These phenomena can become a great concern as a potential health hazard. In order to help control hazards, safety guidelines have been put into place to protect and limit exposure to members of the public and the workforce from unhealthy levels of radiation. Radiation detectors aid in this task by monitoring areas and evaluating exposures to personnel. Monitoring areas where there is a potential for radiation exposure is critical in the evaluation and protection of people that have access to these areas. Limiting exposure is an essential requirement for radiation protection (a principle known as ALARA – As Low As Reasonably Achievable); in order to achieve this health physicist must be able to depend upon reliable detectors.

Lineal Energy Transfer (LET)

Dosimetry can only be as reliable as the detectors that collect the data; therefore, an appropriate method for characterizing the events that take place due to radiation is needed. The importance of describing these events was recognized when early observations showed that equal amounts of energy deposited by some radiations were more damaging than others. For example, alpha particles and neutrons are more biologically effective than beta particles and gamma rays. This led to the concept of linear energy transfer. Linear energy transfer (LET) is important when quantifying the amount of energy imparted in a small site size (in the microscopic range) and determining a dose. LET can be described as the rate of energy lost by ionizing particles as they traverse a medium (Cember 1996). High and low LET particles interact differently in matter. Gamma rays (photons) are considered to be low LET radiation because their charged secondaries are electrons that have low energy transfer. These secondaries are produced in three different ways either by compton scattering, photoelectric effect or pair production. The material composition and photon energy before the interaction will determine the probability of each process occurring. The resulting electrons interact by electronic repulsion (columbic forces) with the electrons of the material irradiated. This process can result in the excitation or ionization of other molecules. High LET radiation (which consists of heavy charged particles like protons, alpha particles and other heavy ions) have characteristically dense linear particle tracks and can produce numerous ionizations along its track. The differences between high and low LET radiation are illustrated in Figure 1.

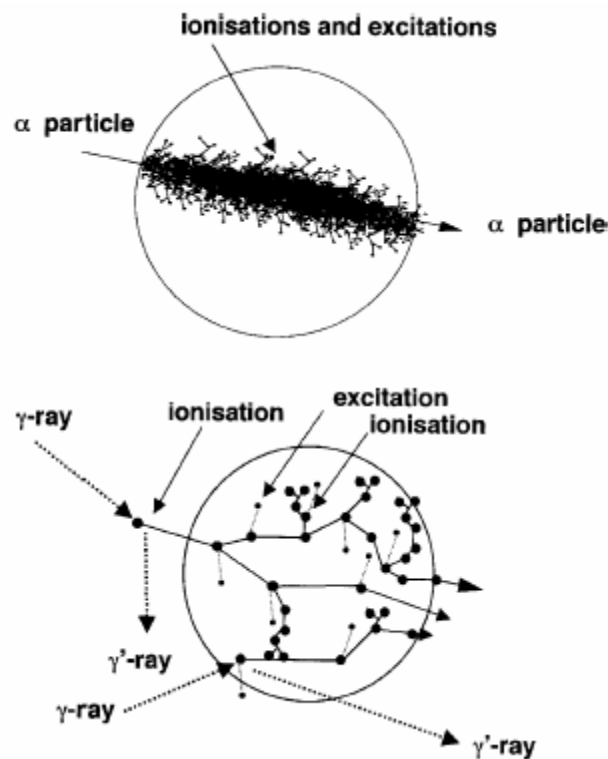


Figure 1. A Representation of High and Low LET Particle Tracks (Pouget and Mather 2001).

These particles deposit energy more rapidly at the ends of their tracks in accordance to Bragg's law. High LET radiation also has a high relative biological effectiveness, RBE, which is the ratio of a standard radiation dose to the dose of radiation of interest that will produce the same biological effect. The RBE of low LET radiation is 1 but high LET radiation can have a RBE of up to 4 for cell killing or 100 for mutation (Pouget and Mather; Rossi and Zaider 1996). This LET/RBE relationship underlines why LET is extremely important in radiation protection, especially high LET radiation.

Microdosimetry

Matter and radiation can be thought of as a mixture or combinations of discrete particles, and because of this discrete nature many different fluctuations in their physical properties can occur. In addition to this there are also many stochastic (random) processes in radiation. These random processes include radiation decay, absorption of photons, scattering of photons, and rate of energy loss of electrons to name a few. These random actions can lead to variations in the energy deposited in identical targets.

Microdosimetry is the systematic study and quantification of the spatial and temporal distribution of absorbed energy in irradiated matter. It aids in the calculation of dose despite the uncertainties from these kinds of random processes (Rossi and Zaider 1996). The detector design being developed in this project will be helpful in microdosimetry because it will allow the determination of doses in small sites where the random nature of radiation can make the determination of a dose difficult.

As mentioned earlier there are many fluctuations when dealing with radiation. Straggling describes the fluctuations in charged particle range and energy loss. Statistical fluctuations can occur in a number of collisions along the path of charged particles and in the amount of energy lost in each collision as the particles cross different materials. As a result, a number of identical particles starting out under identical conditions will show a distribution of energies as they pass a given depth, and a distribution of pathlengths traversed before they stop. Energy straggling is the phenomenon of unequal energy loss under identical conditions and the existence of different pathlengths is

referred to as range straggling (Turner 1995). These processes further illustrate that charged particles lose their energy discontinuously.

Radiation can also be thought about in terms of relevant (energy) transfer points. These transfer points occur where incoming radiation imparts a fraction of its energy to matter. If the transfer of energy is greater than the minimum energy required for an ionization, it is defined as a significant energy deposit and a significant transfer point. Therefore, if no ionizations occur then the transfer points are not significant. Ionizing radiation can also differ in effectiveness because of the local concentrations of absorbed energy. Local concentrations of absorbed energy are governed by the rate of energy loss in charged particle tracks. LET has been considered to be the quantity to which the difference in effectiveness is attributed (Rossi and Zaider 1996).

LET can be used to estimate the amount of energy absorbed when radiation passes through a specified target, but some simplifications have to be assumed. These assumptions can be seen in Table 1.

Table 1. LET Assumptions

Assumption 1	Energy loss is continuous and constant.	In reality this does not happen.
Assumption 2	Energy loss is confined to the track.	In reality this is invalid because of delta rays but using a delta cut off improves the estimate.
Assumption 3	Particle range is large compared to dimensions of the site.	This does not hold for low energy radiation.
Assumption 4	Particle trajectories are rectilinear (moves in a straight line).	This may not be so due to track curvature so more energy is deposited than assumed by the rectilinear track.

LET can only be used to characterize the mean energy deposited in sites because of varying track lengths intercepted by the site. Because LET can only describe mean energy the only assumption that could be made would be related to the probability of effects, but most radiation effect probabilities are not proportional to the absorbed dose. Consequently, LET is not very effective for estimating biological effectiveness.

Determining dose and other important quantities involves taking an average measurement of energy deposited. However in ionizing radiation a single event can cause cell death or can lead to the creation of chemical species that can degrade molecules (like DNA). Therefore using average quantities to determine dose would be appropriate for large site sizes but can result in large errors for small sites that can be of biological importance. Even though there are some different properties behind radiation detection in micro and macroscopic sites, there are common properties that they both share.

Radiation Detection

Detectors in general work when radiation interacts with some active surface or volume of the detector. These interactions then produce charged particles (or ions) that can be collected and/or counted (directly or indirectly). The number of secondary charged particles can then be related to the original incoming radiation and its energy. This allows radiation exposures and doses to be determined. A detector's resolution is important when collecting this information and determining the dose due to radiation. In

addition to this, the detector calibration further assists in determining how much energy is deposited in a detector volume.

Resolution is important in radiation detection because it limits the measurement of small differences in the energy deposition spectra produced by radiation. A detector with poor resolution produces characteristic broad peaks due to an uncertainty in the conversion of deposited energy to output pulse height, while a detector with good resolution will have sharp peaks. Resolution is related to the electron-volt per ion pair, W , of the detection material (gas or solid material). The W value determines the number of ionizations that are produced per unit energy deposited. Because production of an ion is a random event the number of ionizations that are initially produced by the detector determines the inherent resolution. A larger number of initial charge carriers will result in a better detector resolution. This is demonstrated in the resolution equation below. Often electronic noise, variation in a proportional counter's gas gain, high voltage drift and etc. result in resolutions that are substantially poorer than this inherent resolution (Knoll 2000).

$$R = \text{FWHM} / H_0 = 2.35 (F / N)^{1/2} \quad (1)$$

FWHM = Full Width Half Maximum

H_0 = Centroid

N = Number of charge carriers

F = Fano factor

Resolution is one property of detection that allows radiation sources to be determined (qualified), but in order to make a comparative analysis (quantify) the detector must be calibrated. Detector calibration will allow the determination of how

much radiation is present. The calibration of proportional counters is aimed at converting the pulse height, h , corresponding to the energy loss of a charged particle and the associated secondary particles crossing the detector into energy deposited (Gerdung et al. 1992). Mono-energetic alpha particles can be used to calibrate detectors. However, because alpha particles are high LET and can deposit large amounts of energy in a relatively small site size, they can cause large dead times that can paralyze the detector. In order to eliminate this problem while still allowing enough particles into the chamber to obtain statistically relevant data a small aperture (collimator) can be used. This will give a standard of energy deposited so that the detector can be used to determine the energy of other unknown radiations.

Detectors

Tissue equivalent proportional counters are gas filled detectors that can be used to simulate interactions and energy transferred to small tissue volumes, which allows the absorbed dose and radiation quality to be determined. These detectors can also be used to measure the dose from neutrons, charged particles and photons. Some common applications for tissue equivalent proportional counters (TEPC) are monitoring in areas where a mixture of neutron and photon radiations may be present and the characterization of radiation in aircraft and in space. Area monitoring with TEPC has been performed at nuclear power plants, nuclear fuel processing plants and particle accelerators used in medicine and physics research (Schuhmacher 1992). These monitors are used to assess the dose and radiation quality to determine the effectiveness of

shielding and to interpret readings of individual personnel dosimeters (Schuhmacher 1992). However, when measurements are made at low dose-rates the TEPC has to be relatively large (typically 12 cm diameter) to obtain adequate statistical precision in a reasonable time. This makes the detector relatively inconvenient to use.

The specific use of a TEPC detector plays an important role in determining the detector design and how it will be used. Since energy deposition in a small volume appears to be a good way to characterize the radiation quality, and because a low pressure proportional counter can simulate a small site size tissue equivalent proportional counters are of importance. Design consideration for detectors in general, and specifically for TEPC, that should be kept in mind will now be discussed. Several factors to consider when building and designing a detector are cost, size, weight, sturdiness, power consumption, ease of operation and real time display of data (Kunz et al. 1990). Ease of construction is an important aspect of building a detector because of its impact on cost. Ease of construction is also related to detector design because the design can limit which parts can be machined. Cylindrical detectors are the simplest to design and construct, but their chord length distribution is more complicated than that of a spherical detector. In order to achieve a less complex chord length distribution the length, ℓ , of the cylinder and the diameter, d , are usually made equal. Other design features of the detector that must be considered are materials, gas gain and the vacuum seals used. It is very important to have a uniform gas gain in a detector. This allows the particles that are formed in the detector walls to produce the same avalanche

independent of the particle position. This is done by keeping the electric field constant along the length of the anode.

Ideal detectors would be portable and easy to handle, large enough to obtain an adequate number of counts in different radiation fields and indicate the radiation quality, if high LET radiation is present. In addition to these properties, detectors that will be used for personnel monitoring should also be constructed of tissue equivalent materials in order to measure the absorbed dose. However, it is easier to construct a gas filled cavity and collect the ions that are produced in the wall by incoming radiation than constructing a solid tissue equivalent proportional counter. In other words, we really want to determine the dose in the detector wall but we need the gas in order to collect and measure the ionizations that take place. In order to make this determination, cavity theory gives the relationship between dose in the detector wall and dose in the detector gas by the use of average stopping power ratios, as long as certain criteria are met. One criteria is that the particle ranges are large compared to the gas volume so that there are no disturbances in the charged particle field. The second criterion is that the absorbed dose deposited in the cavity results only from the charged particles that cross it. This is solved by creating a small detector cavity. Tissue equivalent proportional counters achieve this by operating at low gas pressures, a fraction of an atmosphere. Because of this TEPC's can be used to determine absorbed dose. A proportional counter that operates under these conditions possesses favorable properties that are useful in radiation detection and personnel monitoring. However there are still modifications and different designs that allow further improvement of detection properties.

Multi-Chamber Proportional Counters

In order to try to attain properties that are close to ideal for routine radiation monitoring, a multi-chamber tissue equivalent proportional design was developed. This design allows an increase in surface area for the same total volume when compared to a standard spherical detector. The surface area of a radiation detector is important in the production of charged particles. The incoming radiation interacting in areas of higher density (detector wall) results in the number of events in the gas filled cavity being proportional to the surface area of the cavity. The multi-chamber design increases the surface to volume ratio by subdividing the active volume of the detector. The theory behind this concept is that the surface area of a cylindrical detector increases approximately with r^2 but the volume is proportional to r^3 . Therefore, if an arrangement of smaller detector volumes is used, then the total surface area can be kept constant while the actual volume of the detector decreases. This advantage can however be reduced by the volume occupied by the material required to separate a specific volume into individual sub volumes (Rossi 1983). Each of these subdivided chamber volumes will act as a small detector whose information can be combined to give data on incoming radiation. This will allow the construction of a detector that has the same counting efficiency as the more commonly used five inch spherical detector but in a more compact design. By using tissue equivalent materials the type of interactions that occur will simulate the same type of interactions in the body. This principle of multi element construction should also be useful for other radiation detectors such as ionization chambers and Geiger counters (Rossi 1983). This type of detector was first constructed

by the Department of Radiation Oncology Columbia University (Rossi 1983). Figure 2 shows a cut away view of this detector. While this detector does have many good features it is not without faults. One drawback to this design is that if a charged particle is able to cross more than one counter septum ionizations will be produced in two volumes at essentially the same time and the result will appear to be an increase in measured LET (Rossi 1996). To prevent this, the chamber walls and septa thickness must be equal to the maximum charged particle range. Another negative aspect of this design is that the electric field at the ends of each cylindrical cavity tends to be non-uniform. This is because the radius of the hole for the anode wire is much smaller than the radius of the rest of the detector wall. To correct this problem and maintain a constant electric field strength along the anode will require corrections for any change in distance between the anode and detector boundary, thus eliminating fringing or edge effects that might occur at the ends of the anode wire (Rossi 1983). The designs proposed in this work will accomplish this through the use of field tubes or guard rings.

Field tubes are small, electrically conductive, tubes that surround the ends of the anode. They are held at a potential relative to the anode wire that is equal to the potential that would exist at the outer radius of the field tube (Braby et al. 1995). The ionized particles that are formed along the length of the field tubes will not reach the anode. So when fringing occurs at the ends of the detector the ions that are produced in that area will be collected on the field tube and will not contribute to the resulting radiation spectra. Field tubes are known to be effective and are used routinely in cylindrical detectors, but they reduce the active volume in the detector. Therefore, the actual

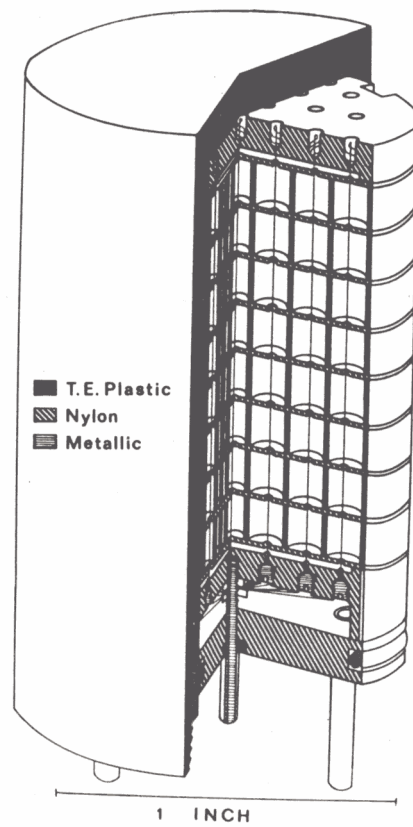


Figure 2. Experimental Multi-Chamber Proportional Counter (Kunz et al. 1990)

detector size will need to include the length of the field tube and the actual counting volume. Sets of guard rings that operate at voltages intermediate between the anode and cathode voltages in order to reduce the detector volume where fringing is significant are an alternative to field tubes. When the detector is powered the intermediate voltage on the rings should correct the distortion in the electric field. These guard rings will only partially compensate for the distortion. The gain will be different in a small volume at the end of the detector because of this. The field tube and guard ring voltage is calculated using the following equation which gives the value of the electric field at radius (r) (Knoll 2000).

$$\xi = V/r \ln(a/b) \quad (2)$$

V = the voltage applied between the anode and cathode

a = the anode wire radius

b = the cathode inner radius

CHAPTER III

METHODS AND MATERIALS

A test detector, consisting of a long cylinder divided into segments using divider partitions with different guard ring configurations (single ring, double ring and field tube) was used to test alternative designs. To construct the detector cavity a 6.35 millimeter ($\frac{1}{4}$ inch) radius, 6.35 millimeter ($\frac{1}{4}$ inch) deep, semi-circular groove was cut into a solid piece of tissue equivalent plastic. This step was repeated so that there were two matching grooves that yield a cylindrical cavity that had a 1.27 centimeter ($\frac{1}{2}$ inch) diameter. Cross grooves were then cut into the plastic to accommodate the counter dividers. The distance between the cross grooves determines the length of each detector. The distances were 1.27 cm ($\frac{1}{2}$ in) and 2.54 cm. (1 in.) for the guard ring and field tube designs respectively. These dimensions were chosen to limit the chord length distribution of the particle tracks. The optimum chord length distribution occurs when the detector length is equal to the detector diameter. For the field tube design, the field tube length should be $\frac{1}{2}$ the detector diameter so the total detector length is twice the diameter. A lip was cut along the edge of each tissue equivalent plastic block to align the two halves and hold them in place. This can be seen in Figure 3.

After the detector body is constructed the field tubes were cut so the exposed length is equal to the radius of the detector. Then, a piece of insulator was placed inside

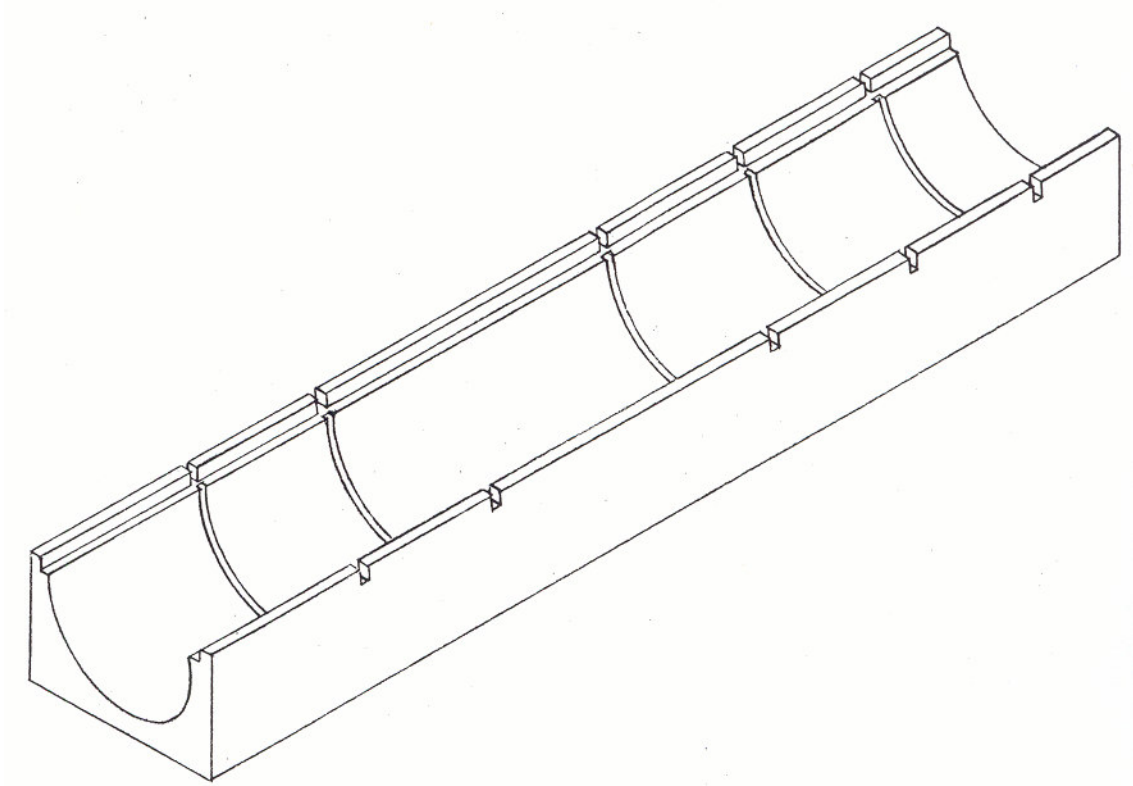


Figure 3. Test Detector Cavity and Surfaces

of the field tube in order to prevent the anode from shorting out. A short will be a serious problem because of the voltage differences between the anode and the tube itself, which will be a fraction of the cathode voltage. Afterwards, the ends of the field tube were spun over in order to keep the plastic insulator inside of the field tube.

Guard rings were developed on a two sided printed circuit board using a photographic resist but first they were designed with the aid of computer drawing software. Three different designs were tested to try to minimize the effects of fringing without unnecessarily increasing the length of the detector. The three designs tested (single ring, double ring and field tube) are pictured in Figure 4.

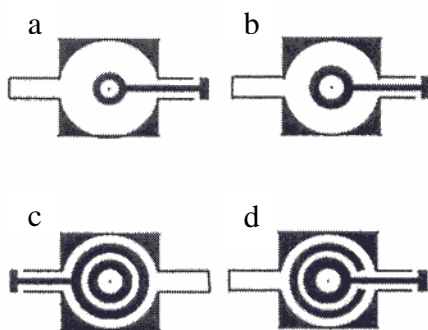


Figure 4. Design of Guard Rings

a. field tube version, **b.** single ring (identical on the back side), **c.** front side of double ring form, **d.** back side of double ring

Each design was produced in a left and right hand version in order to provide the necessary connections to the voltage source. The rings are also double sided in order to make efficient use of the volume in the detector cavity. The double ring configuration was designed with a connection tab for the outer ring on the front side and the inner ring connection is on the back side. This detail is illustrated in Figure 4. When the drawings

were finished positives were printed on transparent plastic film using a laser printer. However because the printer cannot produce a true black image, it was necessary to use a stack of three copies of the design so the image could be properly developed. The masks were carefully aligned on the photo-resist coated circuit board and exposed to an intense light for 2 minutes. After this the board is placed in a developer solution and quickly removed seconds later. This is done to produce an acid resistant image on the copper surface of the circuit board. Next, the board is placed in an acid bath to be etched. During this process the acid etches off all of the copper that was previously exposed to the light. This process takes 15 to 20 minutes. The board is then rinsed and placed in acetone to remove the developed resist from the copper that was not exposed to light. Then, the board is rinsed again, tin plated and then thoroughly washed in water. Finally, the individual pieces were cut out and trimmed to size.

In order to evaluate the guard ring design, the pulse heights produced by collimated beams of alpha particles crossing the detector at different distances from the end were compared. To collimate the alpha particles small holes were drilled on the center line of the detector at specified distances from the divider/detector end. The collimator hole size is important because if it is too small the count rate will be very low and will result in extremely long counting times. If the holes are too large then too many particles will enter the chamber resulting in high dead times and possibly paralyze the detector. To determine the hole size reasonable assumptions were made about the geometry and source activity. The possible hole sizes were then tested in unused areas of the detector. The optimum collimator size was determined to be 0.813 mm. A hole was

placed at the detector's center, 3.969 mm from the center (end position) and 3.175 mm from the center (off-center position).

After the collimator holes were drilled the chamber dividers were installed. First, the field tubes are placed in their proper ring and soldered to make an electrical connection. Now the detector was assembled with the guard rings placed in the cross grooves in the tissue equivalent plastic cut earlier. Figure 5 shows an assembly drawing of the detector cavity to better illustrate this. Then, a single 0.0254 mm (one milli inch) diameter anode is drawn through the center of all of the counters. To do this the anode wire is soldered to a piece of piano wire to aid in threading it through the center of the detector dividers. Afterwards, a stand off is secured in a tapped hole at both ends of the detector array. Then extended wire hooks made of piano wire are soldered to the stand offs. Next, the ends of the anode were soldered to the hook using a 71% ZnCl_2 /29% NH_3Cl soldering flux. Figure 6 shows this. This design positions the chamber dividers and the mounts for both ends of the anode on one piece of tissue equivalent plastic. This allows the detector cavity to be easily accessed for repairs or to make changes to the detector without having to disassemble connections to the guard rings. Next, electrical connections for the guard rings were completed and the voltage divider was assembled. The proper voltages were calculated by using Equation 3 stated earlier (the appropriate radii is listed in Table 2). The result can also be seen in Table 1 and the schematic for the voltage divider is shown in Figure 7.

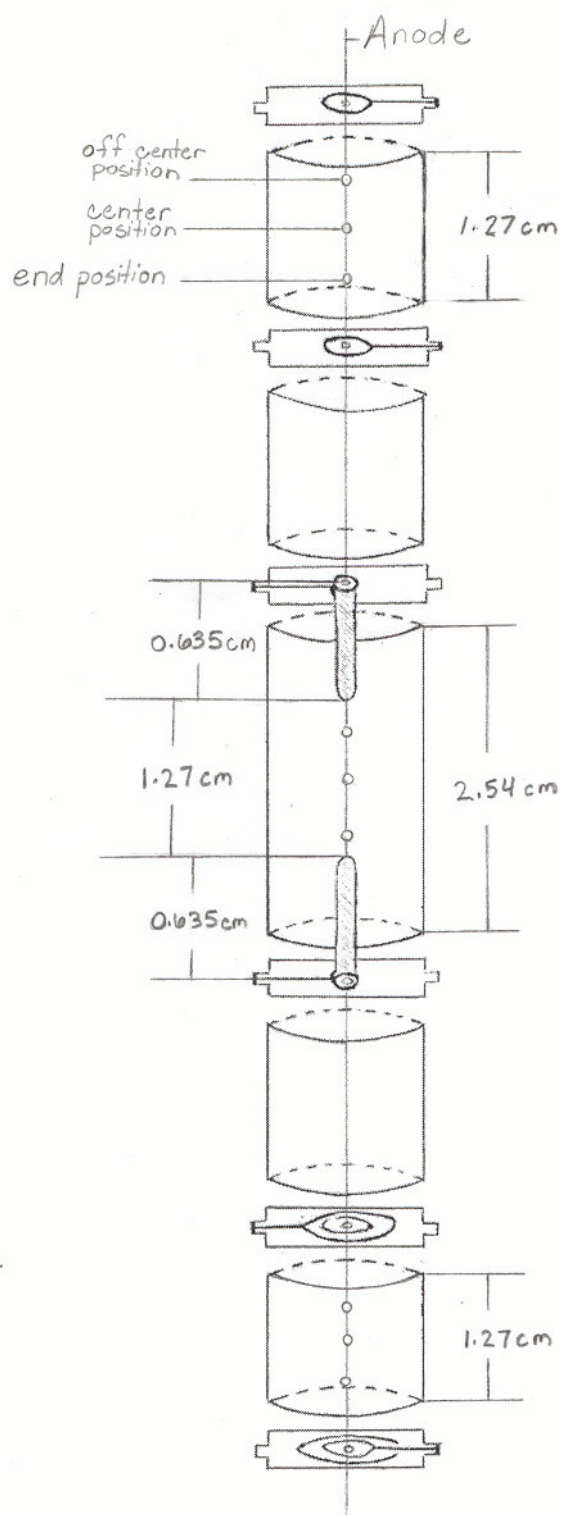


Figure 5. Assembly View of Detector Cavity

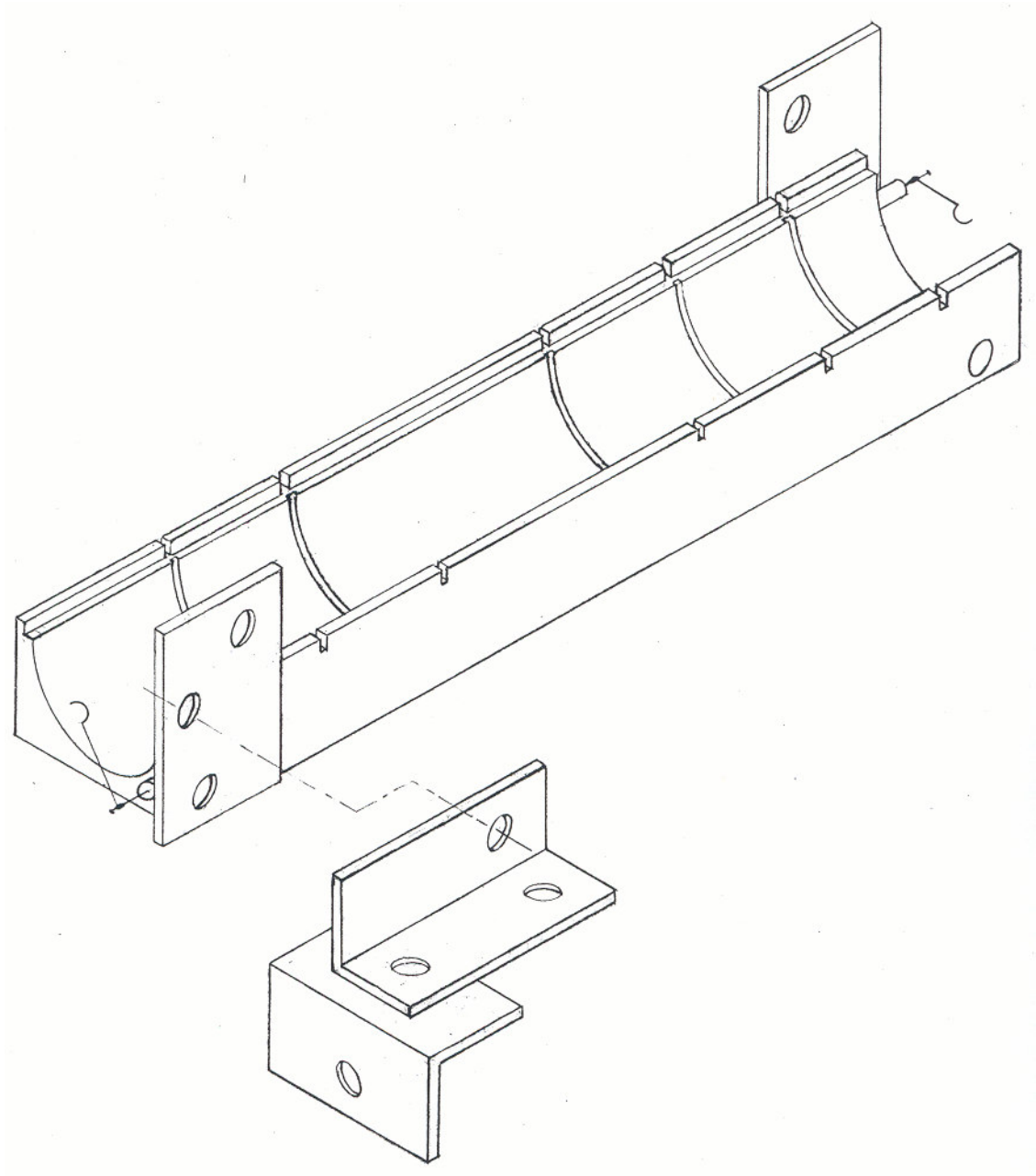


Figure 6. Detector Cavity and Attachments

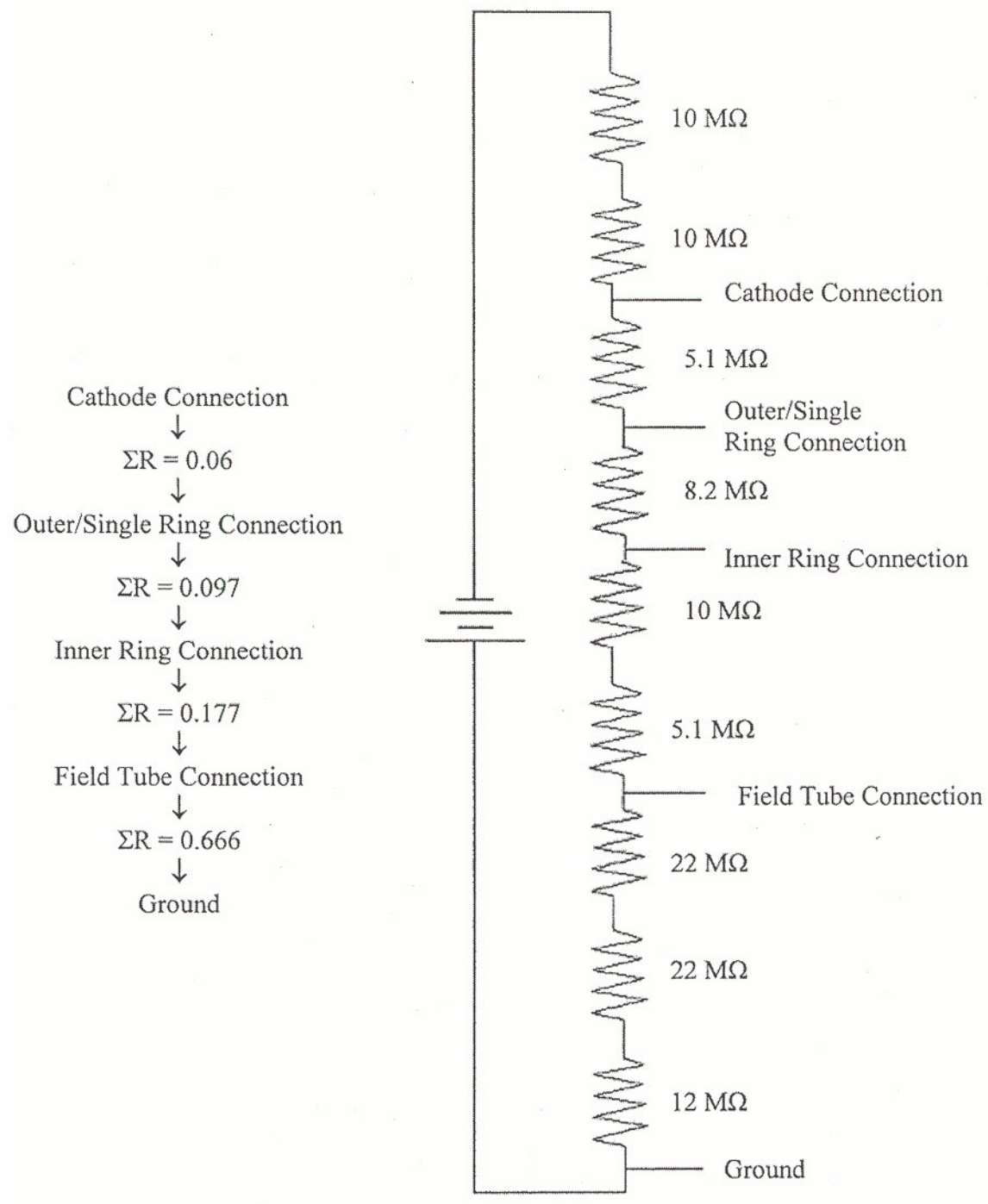


Figure 7. Design of the Voltage Divider Circuit

Table 2. Radii and Voltage Ratios

Configuration	r (mm)	$\epsilon : V$
Field Tube	0.7938	1 : 0.66539
Single or Inner Ring	2.381	1 : 0.84217
Outer Ring	4.366	1 : 0.93971
Cathode	6.35	1 : 1
Anode	0.0127	ground

These radii were taken at the approximate center of the rings

Afterwards, a small piece of adhesive coated plastic sheet (the kind used for document laminating) was placed between the detector body and each one of the tabs on the guard ring in order to prevent a short circuit between the two. Then the detector was fitted with a Lucite clamp to help hold the two halves of the detector together and an extension bracket to position the detector so that it can be tested. This is shown in Figure 6. After the detector is properly set up it is held in a vacuum for about 2 -3 days to get rid of the impurities (water from the atmosphere) that have adsorbed to the plastic. Then the vacuum chamber is flushed with propane and pumped to the pressure that simulates a $2\mu\text{m}$ site size. Finally, each design was evaluated by taking 30 min. counts and collecting a spectrum for each position for each detector.

CHAPTER IV

RESULTS AND DISCUSSION

Each spectrum was evaluated and the resolution was determined. Table 3 shows the results below.

Table 3. Resolutions

Resolution Calculations				
Detector Configuration	Centroid	Channel Width	FWHM	% Resolution
Double ring end	139	153 – 128	25	17.985612
Double ring center	103	109 – 99	10	9.708738
Double ring off center	104	110 – 99	11	10.576923
Field tube end	109	114 – 96	18	16.513761
Field tube center	97	103 – 93	10	10.309278
Field tube off center	99	101 – 93	8	8.080808
Single ring end	155	178 – 145	33	21.290323
Single ring center	117	125 – 112	13	11.111111
Single ring off center	141	148 – 121	27	19.148936

This data is the result from collecting a spectrum at each position for 30 minutes (See Appendix). Analysis shows that the field tube configuration produces the detector with the best resolution. Data for the double ring and field tube setups show a difference in the gain between the center/off center positions and the end position. This is determined by the centroid shift. This is also true for the single ring design. This indicates that there is still some fringing at the ends of the detector. The detectors' resolution also decreases while moving toward the detector end, for the single and double ring configurations. The

field tube did not follow this pattern; this possibly occurred from a non uniformity in the anode at the end of this detector volume or poor statistics. The poor statistics could have resulted from a small uncertainty in the source positioning which resulted in a lower counting rate. Overall the resolution is within 2% for the center position and 2 – 10% for the off center position. The resolution at the end of the detectors is within 5% of each other. This shows that at this position that all of the designs perform about the same at the end of the detector.

CHAPTER V

CONCLUSIONS

The positive results from this project prove that it is feasible to build a large scale version of a multi-chamber proportional counter with either single or double rings. The field tube design does have good resolution but the detector is two times longer than the other two designs. This will make a full scale version of this detector a lot larger for a 1 - 2% (double ring) or 1 - 5% (single ring) gain in resolution. The surface area of a 12.5 cm (5 in.) spherical detector is 506.71 cm^2 so it can be replaced by 67, 1.27 cm x 1.27 cm ($\frac{1}{2}$ in. x $\frac{1}{2}$ in.) volumes. Figure 8 shows a possible design for this type of detector and Figure 9 shows what the series of subdivided volumes may look like. The 67, 1.27 cm ($\frac{1}{2}$ in.) detector array can be fitted in a single layer in a plastic disk with same outside diameter as the spherical detector. Since a single plane of detectors would be only about 1.91 cm ($\frac{3}{4}$ in.) thick, rather than the 16.51 cm (6.5 in.) required for a spherical detector with its anode supports and insulators, the gross volume of the detector can be reduced to about 15% of that of the spherical detector.

Some modifications may further improve the resolution achieved with guard rings. One possibility is optimizing the radius (finding the radius and width of the ring that will minimize the fringing and result in the best possible resolution). Another possibility for improvement is researching the best position on the ring to calculate the electric field (for this project the radius was taken at the center of the ring). This may

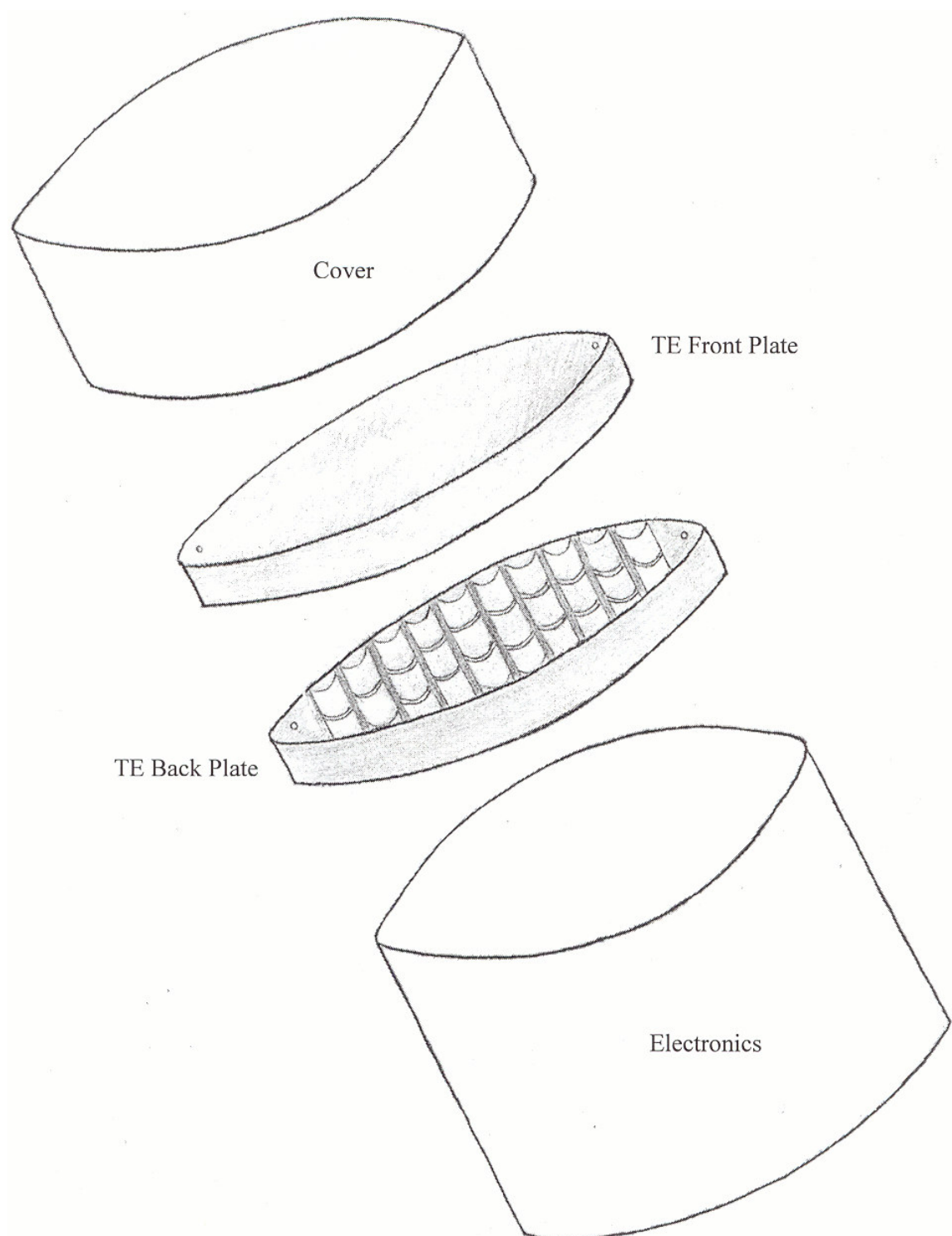


Figure 8. Exploded View of Proposed Multi-Chamber Tissue Equivalent Proportional Counter

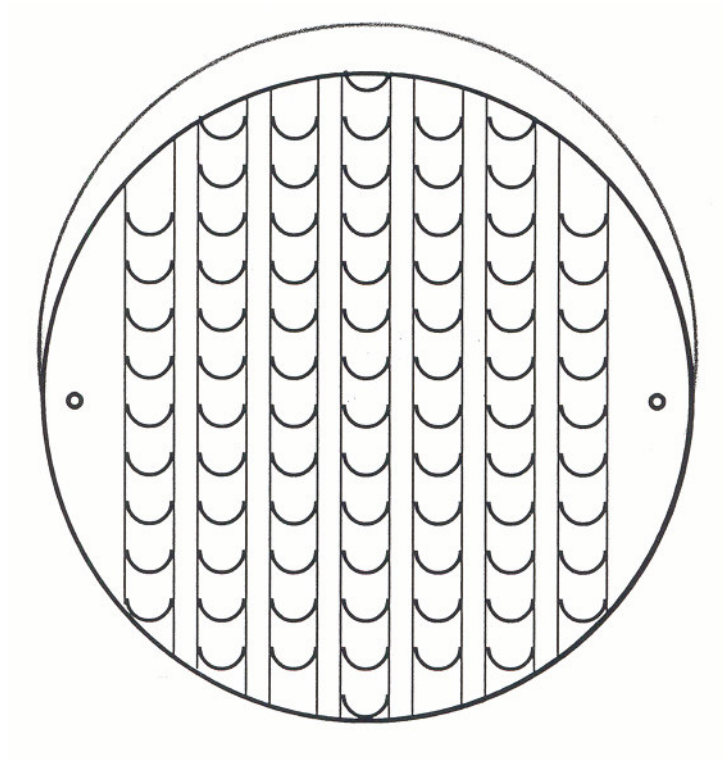


Figure 9. Top View of Tissue Equivalent Plate

change the electric field and therefore the voltage can change to further improve the correcting properties of the rings.

In conclusion, a detector of this type should provide a better way of determining the dose equivalent. It will also have better detection properties. One advantage this type of detector can provide is an increase in counting statistics relative to a smaller detector. Another advantage that should be seen is an increase in detector resolution. Several subdivided volumes can be lined up in series in order to duplicate the surface area of a larger detector and get good statistics in a more compact package.

REFERENCES

- Braby L, Johnson G, Barthe J. Practical considerations in the design and construction of tissue-equivalent proportional counters. *Radiation Protection Dosimetry* 61: 351-379; 1995.
- Cember, H, Introduction to health physics, 3rd ed. New York: McGraw Hill; 1996.
- Gerdung S, Pihet P, Gringborg J, Rossi H, Schrewe U, Schuhmacher H. Operation and application of tissue equivalent proportional counters. *Radiation Protection Dosimetry* 61: 381-404; 1992
- Knoll, G, Radiation detection and measurement, 3rd ed. New York: John Wiley and Sons; 2000.
- Kunz A, Pihet P, Arend E, Menzel H. An easy to operate portable pulse-height analysis system for area monitoring with TEPC in radiation protection. *Nuclear Instruments and Methods in Physics Research A* 299 696-701; 1990.
- Pouget J, Mather S. General aspects of the cellular response to low- and high- LET radiation. *European Journal of Nuclear Medicine* 28: 541-561; 2001.
- Rossi H. Notes: multi-element dosimeters for radiation protection measurements. *Health Physics* 44: 403-405; 1983.
- Rossi H, Zaider M, Microdosimetry and its applications, New York: Springer – Verlag Telos; 1996.

- Schuhmacher H. Tissue equivalent proportional counters in radiation protection dosimetry: expectations and present state. *Radiation Protection Dosimetry* 44: 199-206; 1992.
- Turner J, *Atoms, radiation, and radiation protection*, 2nd ed. New York: John Wiley & Sons Inc; 1995.

APPENDIX

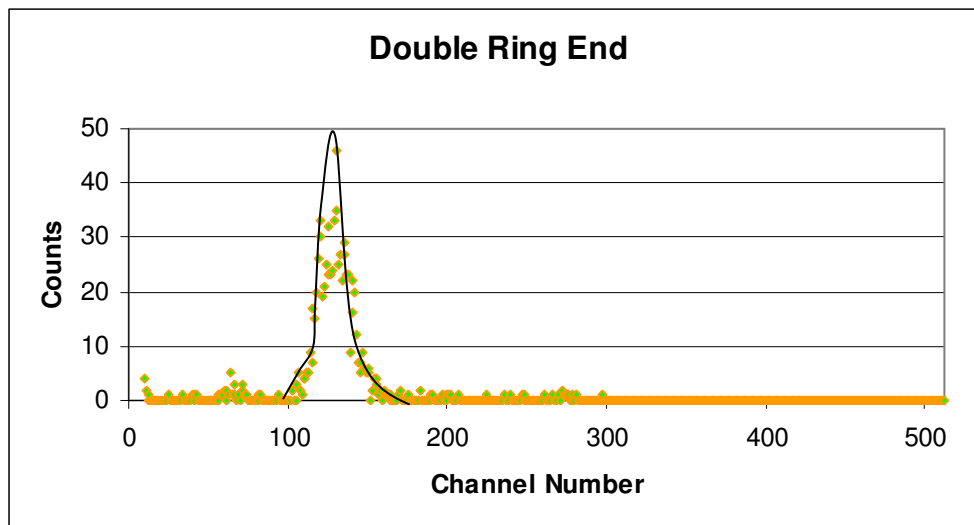


Figure A-1. Spectrum resulting from a 30 min. count at the end position of the double ring configuration.

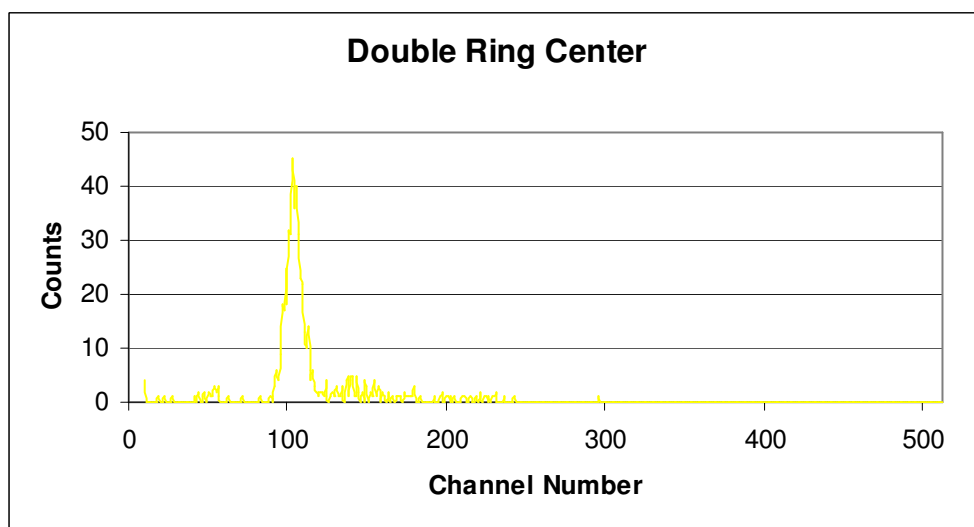


Figure A-2. Spectrum resulting from a 30 min. count at the center position of the double ring configuration

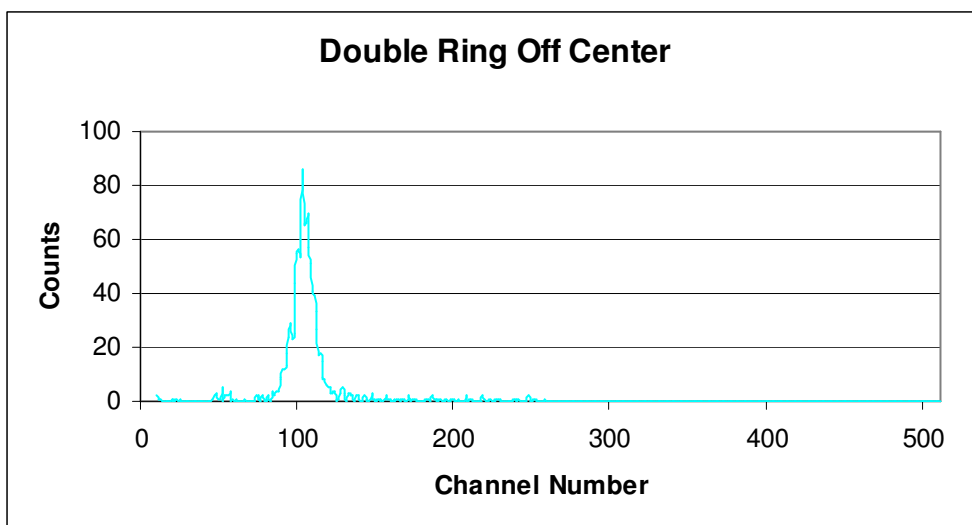


Figure A-3. Spectrum resulting from a 30 min. count at the off center position of the double ring configuration

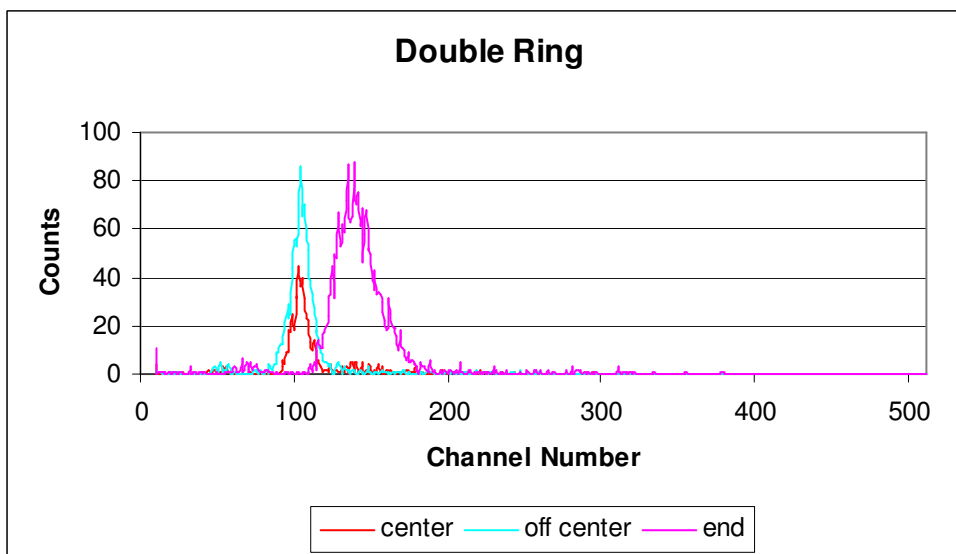


Figure A-4. Overlapping double ring spectra comparing the gain at the different positions

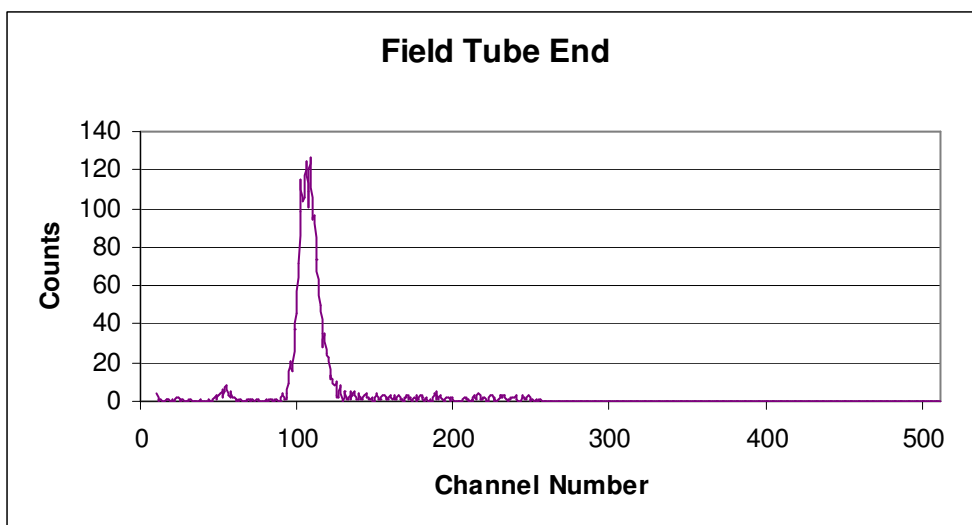


Figure A-5. Spectrum resulting from a 30 min. count at the end position of the field tube configuration

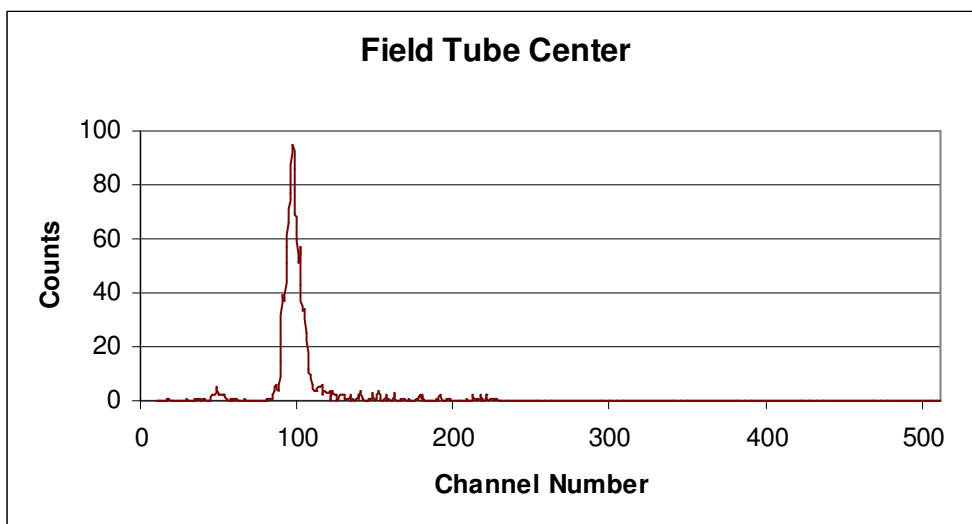


Figure A-6. Spectrum resulting from a 30 min. count at the center position of the field tube configuration

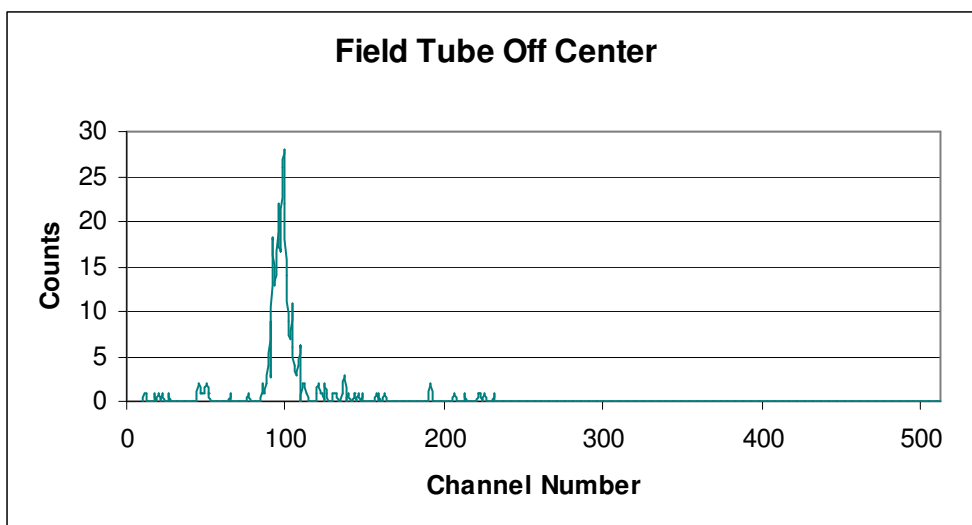


Figure A-7. Spectrum resulting from a 30 min. count at the off center position of the field tube configuration

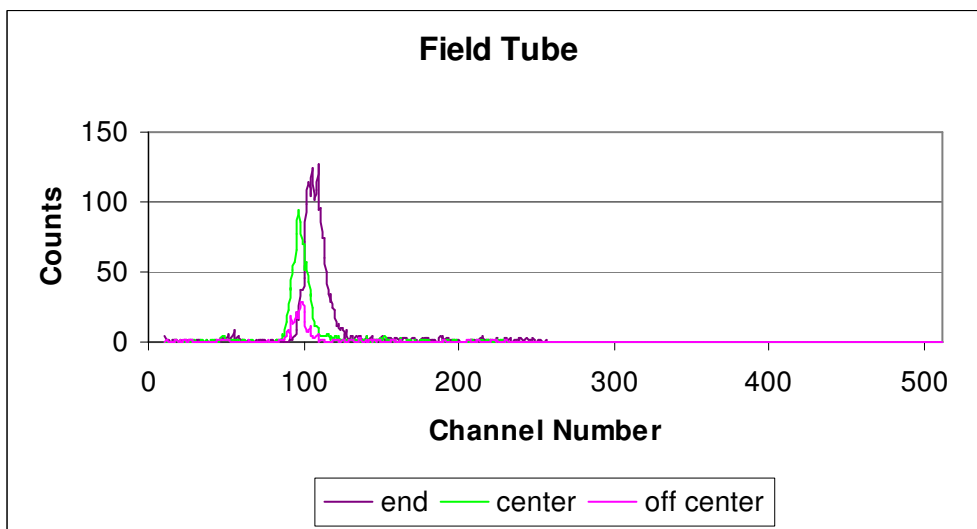


Figure A-8. Overlapping field tube spectra comparing the gain at the different positions

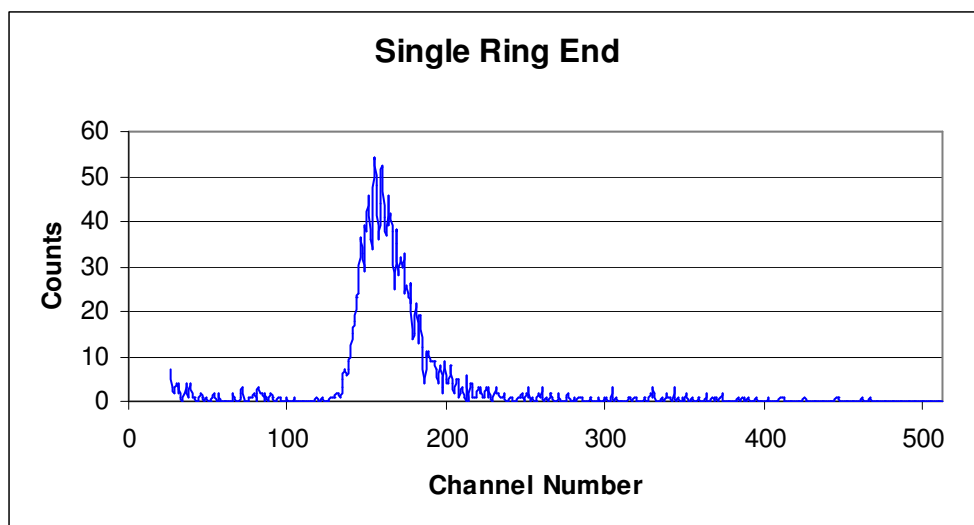


Figure A-9. Spectrum resulting from a 30 min. count at the end position of the single ring configuration

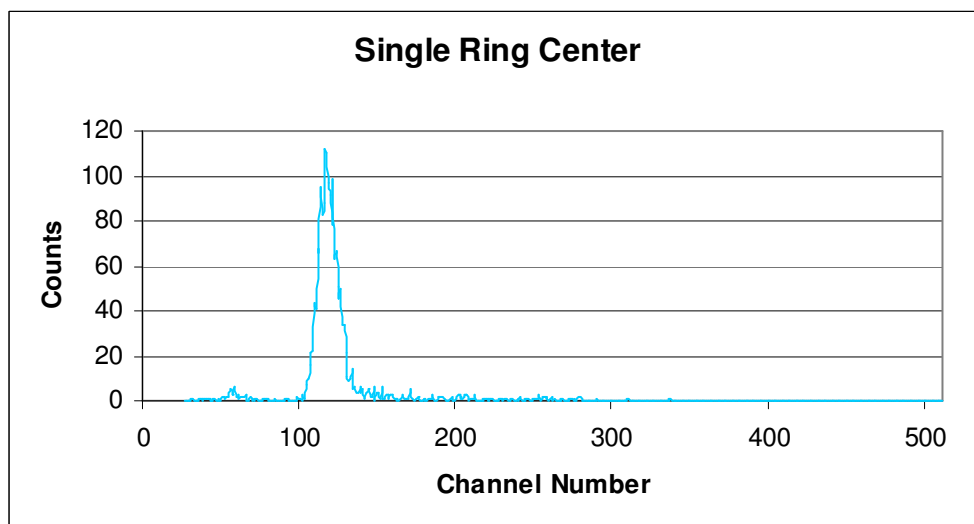


Figure A-10. Spectrum resulting from a 30 min. count at the center position of the single ring configuration

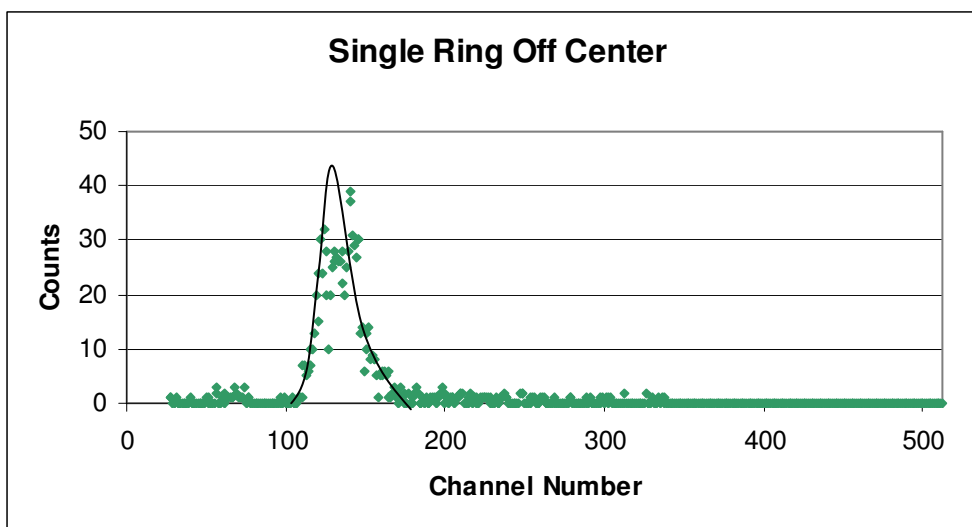


Figure A-11. Spectrum resulting from a 30 min. count at the off center position of the single ring configuration

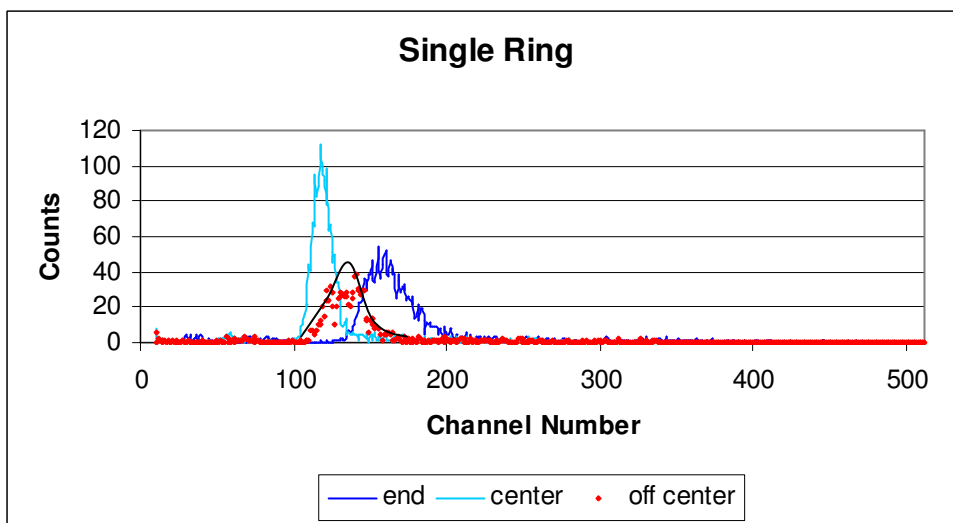


Figure A-12. Overlapping single ring spectra comparing the gain at the different positions

VITA

Temeka Taplin received her Bachelor of Science degree in chemistry from Southern University at Baton Rouge in December 2000. She entered the health physics program at Texas A&M University in January 2003 and received her Master of Science degree in December 2005. Miss Taplin may be reached at the Nuclear Engineering Department Texas A&M University 3133 TAMU College Station, TX 77843-3133. Her e-mail address is ttaplin@hotmail.com.